ANALYSIS OF EFFECTS OF NONUNIFORM BEAM PROFILE ON DOPPLER POWER SPECTRUM AND MEAN VELOCITY

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Abstract: In this paper we present a detailed simulation method to estimate the accuracy of Doppler power spectrum and mean blood velocity, using real CW Doppler transducers with twin-crystal arrangement. The method is based on dividing the sample volume into small cells using the statistics of the total power of Doppler spectrum with the same Doppler shift frequency, and from total power, which predicts errors in the mean blood velocity. Results show that the Doppler angle, vessel depth, and sample volume length are not sensitive functions for both the Doppler power spectrum and error in the mean blood velocity. However vessel diameter, displacement distance and blood velocity profiles have significant effects. Finally, comparisons between simulation and experiment results illustrated a good agreement for parabolic flow profile. These results will contribute to an improved understanding of Doppler power spectrum and error in mean blood velocity in medical ultrasound diagnostics.

Keywords-**continuous-wave, semicircle transducer, Doppler power spectrum, blood velocity, spectral broadening**

I. INTRODUCTION

Ultrasound Doppler is widely used to measure blood flow in the vessel by analyzing Doppler power spectrum for the diagnostic assessment of cardiovascular disease [1,2]. Recently, of particular interest to us is the application of this technique to human subjects during exercise. A telemetry system for measuring blood velocity in exercising subjects by using continues wave (CW) ultrasonic Doppler method has been developed in our laboratory [3] and used for assessing blood flow changes during various physical exercises [4]. Especially, two semicircular transducers were chosen to use in the telemetry system, which is generally applied for a typical clinical CW probe. However, for accurately evaluating blood velocity, it is necessary to measure blood velocity profile, mean velocity and volumetric flow. There was an unsolved problem that remained for the effect of semicircle transducer in the theoretical simulation study, which would cause distortion of the Doppler spectrum and errors in mean velocity in CW system Ultrasonic Doppler that has not been considered. Cobblod and Bascom [5] have provided some numerical calculation methods, but they assumed the transducer to be coincident with transmission and reception.

The simulation studies for Doppler power spectrum have been performed by a number of investigators. Cobblod and Bascom have analyzed the effects of beam profile, insonation degree and spectral broadening on the CW Doppler ultrasound spectrum and mean velocity. Mo and Cobblod, and shung's group have developed the effect of the scatterer's distribution in blood. Consequently, there is a need to understand how both acoustical and physiological factors such as transmitting and receiving Doppler angle and distance, beam profile form, different transducer shape, spectral broadening and scatterer's distribution, affect the distortion of the Doppler spectrum and error in mean velocity on the CW system Doppler.

Thus, the purpose of our study is to extend the theories of above investigators to present a detailed simulation model for accurately evaluating Doppler power spectrum and error in mean velocity using real CW Doppler transducers with twin-crystal arrangement. Finally, to ensure a demonstration of the efficacy of our method, the computed power spectrum was compared to experimental results obtained for different tube diameter and flow velocity with parabolic profile.

II. MATERIAL AND METHODS

A. Theoretical Basis of Simulation Model

Fig. 1 shows transducers, a blood vessel and their respective coordinate systems. It consists of two semicircle transducers at an angle θ to a blood vessel. To simplify the analysis axisymmetric flow inside a cylindrical pipe paralleled to the vessel axis and the skin surface is assumed. The angles α and β between the direction of flow velocity at the point Q and the transmitting and receiving beam, respectively are given by:

$$\cos\alpha = \frac{\mathbf{q_1}\cdot\mathbf{v}}{|\mathbf{q_1}||\mathbf{v}|} \qquad (1)$$

$$\cos\beta = \frac{\mathbf{q_2}\cdot\mathbf{v}}{|\mathbf{q_2}||\mathbf{v}|} \qquad (2)$$

where the unit velocity vector $\mathbf{v} = (0, \sin\theta, \cos\theta)$, vector $\mathbf{q_1} = \overrightarrow{QO_{11}} = (x - x_{11}, y - y_{11}, z - z_{11})$ and vector $\mathbf{q_2} = QO_{12} = (x - x_{12}, y - y_{12}, z - z_{12})$. The angle θ_1 between the transmitting and receiving beams at the point Q can be written as:

$$\cos\theta_1 = \frac{\mathbf{q_1}\cdot\mathbf{q_2}}{|\mathbf{q_1}||\mathbf{q_2}|} \qquad (3)$$

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The spatial axisymmetric velocity profile can be written as the following equation [6]:

$$v(r) = v_{\max}\left(1 - \left(\tfrac{r}{R}\right)^n\right) \qquad (4)$$

where $v(r)$ is the velocity, r is the distance from the center of the vessel of radius R, and v_{max} is the axial flow velocity. A parabolic velocity profile is described when $n=2$. Increasing the value of n makes the profile more blunt until plug flow is described when $n=\infty$.

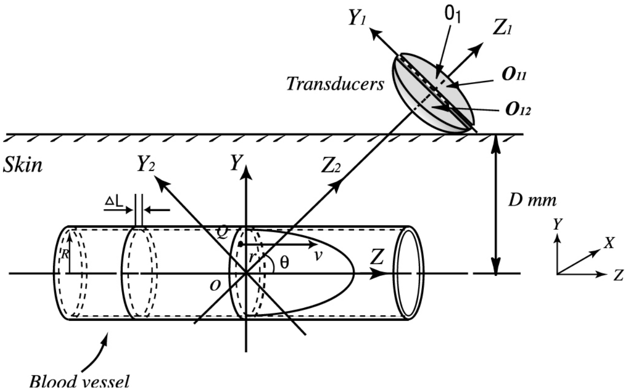

Fig.1 Cartesian coordinate system used to numerically estimate the Doppler spectrum.

B. Calculation of Total Received Power

A semicircle transducer of 15mm in diameter $(2R_b=15mm)$ for emitting ultrasound to the blood vessel and a same semicircle receiver of in diameter for receiving the reflected ultrasound containing Doppler shift signals. Considering the depth of the vessels and the size of transducers, 2.0MHz was chosen as the emitting frequency. Acoustic pressure field in the blood vessel has been calculated by dividing a transmitter into segments and accumulating signals reflected from every part [7]. The total pressure at point Q can be written as

$$P_{inc}(r_1) = j\rho ck \frac{\dot{U}}{2\pi} \iint_{S_1} \frac{e^{jkr_1}}{r_1} dS \qquad (5)$$

where r_1 is the distance from dS to observation point Q, ρ and c are medium density and velocity of sound, $k=2\pi/\lambda$, λ is the ultrasonic wavelength, $\dot{U}$ is the local velocity of the transmitter surface S_1. The particles of radius a suspending in a medium with different acoustic properties are much smaller than the incident ultrasonic wavelength λ ($a \ll \lambda$). The scattering of waves by the particles is usually known as "Rayleigh scattering". The scattering cross section σ relates to the energy scattered by the incident intensity. For Doppler shift frequency f_d the scattered pressure $P_R(r_1, r_2)$ of red cell exerting on the receiving transducer surface can be written as [7]:

$$P_R(r_1, r_2) = -\sigma \cdot P_{inc}(r_1)\, j\rho ck \frac{\dot{U}}{2\pi} \iint_{S_2} \frac{e^{jkr_2}}{r_2} dS \qquad (6)$$

$$f_d = \frac{\cos\alpha + \cos\beta}{c - v\cos\alpha} v f_0 \qquad (7)$$

where r_2 is the distance from an erythrocyte on the point Q to the receiving transducer elemental surface area, and dS defines integration over the receiving transducer surface S_2 area.

The spectrum of the scattered echo of a single particle can be replaced by the voltage at the transducer $V(f_d)$. This voltage is related to the pressure $P_R(r_1, r_2)$ at the transducer by the system response. For the scattering configuration where the transmitting transducer is the same as the receiving transducer and the scattering pressure $P_R(r_1, r_2)$ at the transducer will be transformed into the voltage at the transducer by the efficiency factor $T(f_d)$ of the transducer at the reception as following [8]:

$$V(f_d) = T(f_d) \cdot P_R(r_1, r_2) \qquad (8)$$

In order that the flow velocity of each cell in the elements is treated homogeneously the sample volume is divided into small sector volume elements.

The power spectrum for a certain Doppler shift frequency f_d can be obtained by accumulating sampling cells corresponding to the same f_d. The total received power $P(f_d)$ for Doppler shift frequency f_d can be calculated:

$$P(f_d) = \iiint_{V_S} |V(f_d)|^2 \cdot N dV \qquad (9)$$

where Vs is the total volume of sampling cells and N is number of red cell of per unit volume.

C. Experiment

A flow phantom system was used to experimentally verify the simulation model. Two *2MHz* semicircle transducers of 15mm diameter were used to transmit and receive signals. The transmitted beam angle θ=45° was set to and the depth of a vessel is D=24mm. The test fluid flows from an upper the reservoir to a lower one through a silicon tube of 4 and 8mm internal diameter immersed in a tank filled with water. The flow velocity can be controlled by change the height of upper reservoir. The Doppler test fluid (model 707-G, Generex) is a dispersion of plastic particles in a glycerine water mixture. The particle size is 30±3μm mean diameter. It was a little bigger than the erythrocytes but much smaller than the wavelengths used in the experiments. mean diameter. It was a little bigger than the erythrocytes but much smaller than the wavelengths used in the experiments. Therefore, a scattering is similar to the blood cells. In order to maintain laminar flow with a mean flow velocity $\bar{v}$ in a tube of radius R with a fluid kinematic viscosity γ, the Reynolds number R_e can be written as:

$$R_e = \frac{2R\bar{v}}{\gamma} \qquad (12)$$

The threshold of R_e is approximately 2000 for a silicon tube. In the experiment the numerical values of the

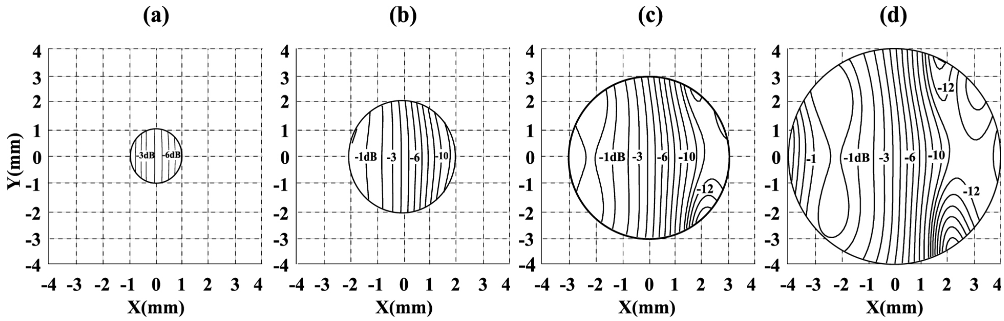

Fig. 2. Ultrasonic pressure fields of the cross-sections of the blood vessel. Ultrasonic pressure fields in isobaric contours across the blood vessels of diameter (a) 2 mm, (b) 4 mm, (c) 6 mm, and (d) 8 mm in 24mm depth from the skin.

parameters are *2R=4 and 8mm*, γ=1.66±0.1 mm^2/s. When v_{max}=70cm/s and 80cm/s, the Reynolds number R_e is 867 and 1960 respectively for the above diameters of tube.

III RESULTS AND DISCUSSION

One reason of spectral distortion arises from the nonuniformity of the acoustic pressure field distribution in the vessel. The acoustic pressure field distribution will vary for different vessel sizes, depending on the extent and position of the region of the beam covering the vessel. Fig.2 (a)-(d) show the multiplying acoustic field for 2-MHz semicircle transducer from vessels of diameters 2.0 mm, 4.0 mm, 6.0 mm and 8.0 mm, respectively, in depth D=24mm. The results show that as increase in area of vessels, the nonuniformity of acoustic field distribution increase.

Fig. 3 shows as sample volume size progressively increases the power spectrum also increases, but error in mean velocity decreases for parabolic flow. When sample volume size is more than the length of intersection of beam and vessel, the distortion of the power spectrum and error in mean velocity are very close to constant.

In Fig. 4 the results show that as the vessel diameter increases both the total received power and spectral distortion increase, in addition, the curves for error in mean blood velocity show increase monotonically, suggesting this significant effect is due to nonuniform beam or incomplete insonation.

Fig. 5 shows the received power spectrum when intrinsic spectral broadening is considered. As velocity parameter n increases the flow profile becomes blunter and the power spectrum approaches a spike at the higher frequencies, but the spectral distortion and error in mean velocity decrease.

Fig. 6(a) shows the effect of beam that displaced from vessel central axis (d_1=0) by successive 0.5mm increments. When the displacement of the beam from d_1=0 causes a fall-off in the power spectrum until at d_1=1.0mm and the Doppler power spectrum distribution becomes close to flat provide beam displacements cause a reduction in the power spectrum. Furthermore, until at d_1=4.0mm, the power spectrum again becomes close to flat. This behaviour arises from beam distribution, which is close to uniform when displacement of beam d_1 is more

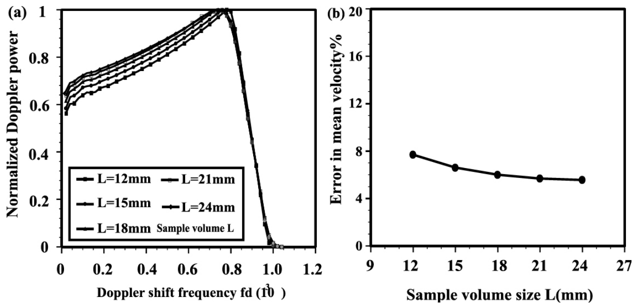

Fig.3. (a) The Doppler power spectrum distributions and (b) errors in the mean blood velocity obtained for sample volume length from 12mm to 24mm for parabolic profiles (R=2mm, D=2.4cm, θ=45^0, v_{max}=50cm/s).

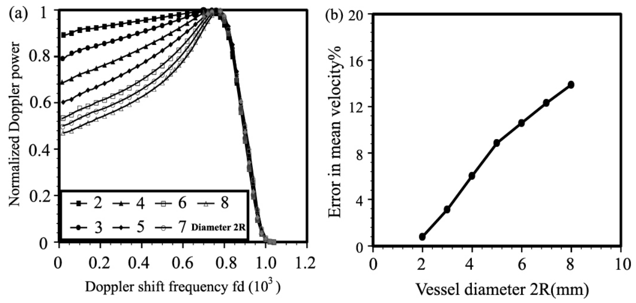

Fig.4. (a) The Doppler power spectrum distributions and (b) errors in the mean blood velocity obtained for vessel diameter from 2mm to 8mm for parabolic profiles (D=2.4cm, θ=45^0, v_{max}=50cm/s, L=21mm).

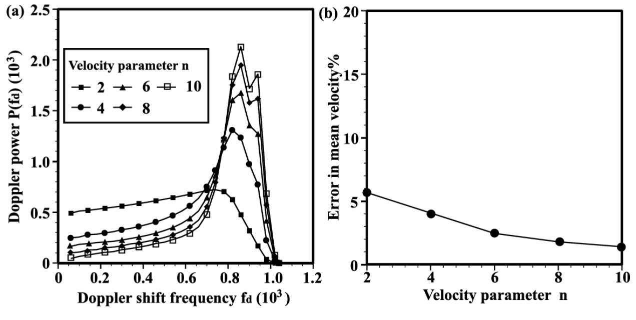

Fig. 5. (a) The Doppler power spectrum distribution (b) errors in the mean blood velocity obtained for different values of the velocity parameter n (R=2mm, D=2.4cm, θ=45^0, v_{max}=50cm/s).

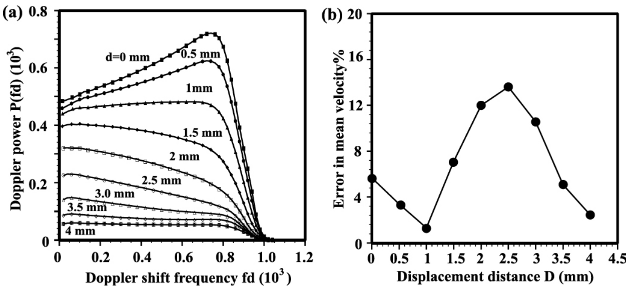

Fig.6. (a) The Doppler power spectrum distributions and (b) errors in the mean blood velocity obtained for displacement distance (d_1) from 0mm to 4mm for parabolic profiles (R=2mm, D=2.4cm, $\theta=45^0$, v_{max}=50cm/s).

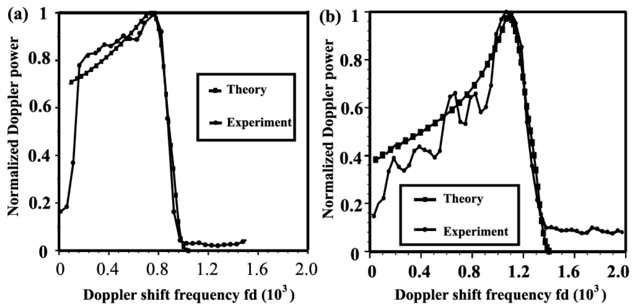

Fig.7. Comparison between simulated and measured Doppler power spectrum for parabolic flow profile with vessel depth D=2.4cm and the 2MHz probe oriented at 45^0: (a) the 4mm ID silicon tube, v_{max}= 70cm/s; (b) the 8mm ID silicon tube, v_{max}= 80cm/s

than 4mm. On the other hand, the effect of beam shape and displacement on the estimated error in mean velocity is shown in Fig. 6. (b). For a semicircular beam the errors are minimum and maximum value at d_1=1mm and d_2=2.5mm. The latter result suggests that the region lying along the beam axis is more heavily weighted than the off-axis regions of vessel, which causes maximum distortion of power spectrum.

Fig.7. (a) and (b) show comparisons between experimental and simulation results for different silicon tube and flow velocity. It can be seen that in each case measured and simulated Doppler power spectrum are given in good agreement for high-frequency end. The lack of power spectrum near the zero frequency in the experimental spectrum may be due to the effects of high pass filter and attenuation.

From the results presented above, it can be seen that the first major contribution in designing our simulation model is that we chose the semicircular transducer for the first time, which is typical clinical CW system Doppler probe with separated twin-crystal arrangement. Thus, our study developed Bascom's work, which assumed coincidence of the transmitting and receiving crystals [6].

The second major contribution is that we considered more sufficiently the factors affected on the Doppler power spectrum and error in the mean blood velocity. With the change of vessel depth, sample volume length, and Doppler angle, the shape of the power spectrum and error in mean velocity remained constant since each component of the spectrum was changed in near the same manner. However, the other factors such as vessel diameter, velocity profile and displacement distance strongly affected power spectrum and error in mean velocity.

IV CONCLUSIONS

This paper has considered a number of factors affected on Doppler power spectrum and error in the mean blood velocity, described a detailed simulation method for calculating the Doppler spectrum for a real CW transducer separated for the transmission and reception, and from the total power to predict the errors in the mean blood velocity. This method makes it possible to estimate the spectral shape for any type transducer. In particular, it was concluded that the Doppler power spectrum and error in the mean blood velocity are significantly affected by vessel diameter, displacement distance and blood velocity profile, less affected by Doppler angle, vessel depth, and sample volume length. In addition, it is also proved to compare computed spectra in good agreement with experiments. We believe our simulation model presented is general and can be applied to the more usual case of spatial spectrum on CW Doppler ultrasonic system.

REFERENCES

1 P. A. Bascom and R. S. Cobbold, "Origin of the Doppler ultrasound spectrum from blood," IEEE Trans Biomed Eng. 43:562-71;1996.

2 P. A. Bascom, K. W. Johnston, R.S. Cobbold, M. Ojha, "Relation of the flow field distal to a moderate stenosis to the Doppler power," Ultrasound Med Biol. 23:25-39, 1997.

3 J. He, Y. Kinouchi, T. Iritani, H. Yamaguchi, H. Miyamoto, "Telemetering blood flow velocity and ECG during exercise," Innov Tech Biol Med. 13:567-77, 1992

4 Y. Hirao, D. L. Zhang, K. Tamukai, Y. Kinouchi, H. Yamaguchi, K. Yoshizaki, "Measurements of maximum blood flow velocities and the propagation time of velocity pulse wave on carotid, brachial and femoral arteries during exercise by an Ultrasonic Doppler method," Technical report of IEICE. MBE99-55, 1999.

5 R. S. Cobbold, P. H. Veltink, K. W. Johnston, "Influence of beam profile and degree of insonation on the CW Doppler spectrum: A theoretical study." IEEE Trans Son Ultrason.; 30:364-70, 1983.

6 Roevros JMJG. "Analog processing of C.W. Doppler flowmeter signals to determine average frequency shift momentaneously without the use of a wave analyzer," Cardiovasular Applications of Ultrasound. 43-54. North-Holland, Amsterdam, 1974.

7 J. Zemanek. Beam behavior within the nearfield of a vibrating piston. J Acoust Soc Am.; 49:181-91. 1971.

8 E. P. Papadakis. Ultrasonic Instruments and Devices. Pennsylvania: New Holland; 1999.